Simulation of Guide Wire Propagation for Minimally Invasive Vascular Interventions

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Abstract. In order to simulate intravascular interventions, a discretized representation of a guide wire is introduced which allows modeling of guide wires with different physical properties. An algorithm for simulating the propagation of a guide wire within a vascular system, on basis of the principle of minimization of energy, has been developed. Both longitudinal translation and rotation are incorporated as possibilities to manipulate the guide wire. The algorithm is based on quasi-static mechanics. Two types of energy are introduced, internal energy related to the bending energy of the guide wire and external energy resulting from the elastic deformation of the vessel wall. Compared to existing work, the novelty of our approach lies in the fact that an analytical solution is achieved. The algorithm is tested on phantom data. Results indicate plausible behavior of the simulation.

1 Introduction

Minimally invasive vascular interventions are performed through catheterization. During these interventions visual feedback is provided by intra–operative X-Ray images. The intervention is performed by manipulating parts of the instruments that are outside the body. In order to successfully carry out these procedures, proper training is required. Providing a training possibility by simulation has a number of attractive properties. No radiation is required and the simulation can be made patient specific.

In order to simulate intravascular interventions, the instruments and the patient need to be modeled. Hereto, tissue types with different elastic properties have to be segmented and modeled. Also the interaction between the instruments and the patient needs to be modeled. Finally, simulation of the visual feedback during the intervention is required.

The simulation of surgical procedures has been the interest of a variety of research groups. Progress in the field of simulating minimally invasive vascular interventions has been made in the last few years [1,4,7,6].

In this paper we focus on modeling the guide wire which is the main instrument used during a minimally invasive vascular intervention. The guide wire allows the radiologist to navigate inside the vasculature.

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To model the behavior of a deformable object, the Finite Element Method (FEM) approach [9] and mass-spring models are both suitable methods. However, mass-spring models are better suited for modeling soft tissue deformation [3]. Furthermore, the FEM approach requires a large amount of computation time, which makes it less appropriate for real time applications. Therefore, we introduce a different approach: an analytical solution of a discretized model of the guide wire. This model contains the required information to add force feedback, supplying a good basis for a realistic simulation of a catheterization procedure.

The outline of the paper is as follows. The aspects concerned with the modeling of the guide wire and the vasculature are discussed in Section 2. Section 3 is dedicated to the relaxation algorithm which determines the guide wire position using energy minimization. The initialization of the relaxation algorithm is the topic of Section 4. Section 5 is dedicated to simulation results on phantom data. Finally, details concerning future research are discussed and some conclusions are drawn in Section 6.

2 Modeling the Guide Wire and Vasculature

Two types of energy are associated with the guide wire. The internal energy of the guide wire that is related to the bending of the guide wire and the external energy of the guide wire related to the elastic deformation of the vessel wall.

The representation of the guide wire and the bending energy associated with the guide wire are discussed in Section 2.1. Subsequently, the modeling of the elastic deformation energy of the vessel wall is discussed in Section 2.2.

2.1 Guide Wire Modeling

The guide wire is modeled by a discrete parametrization. A set of equally large segments connected at so called joints represents the guide wire. The segments are straight, not bendable or compressible and have a fixed length (λ). The segments have to be small to approximate reality.

The guide wire can be represented by storing an array with the 3D position vectors of the joints: $\overline{x}_0, \ldots, \overline{x}_k$. The joint position closest to the insertion tube is called $\overline{x}_0$ (Figure 1). When a new part of the guide wire is inserted into the vessel, the current representation is adapted by adding segments and joints and by computing a new guide wire configuration with the relaxation algorithm.

Bending energy is associated with the angular orientation between two segments. The bending energy increases if the difference in angular orientation between the two segments increases. Bending energy (U_b) per joint is given by:

$$U_b(\overline{x}_i) = \frac{1}{2} c_i {\theta_i}^2 \tag{1}$$

Where θ_i is the angle between two segments connected by joint i (Figure 1) and c_i is a spring constant. This spring constant is related to the stiffness of the joint.

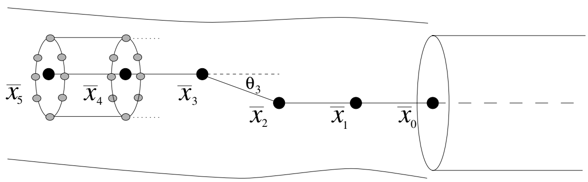

Fig. 1. Representation of a guide wire defined by joint positions $\overline{x}_0, \ldots, \overline{x}_5$. The angle ($\theta$) between two segments as used in equation 1 is illustrated for joint 3. For joint 4 and 5 the sample points on the outer hull of the guide wire, which allows the modeling of guide wires with different thickness, are drawn.

By storing a separate spring constant for each joint, guide wires with different physical properties, i.e. guide wires with a variable flexibility along the guide wire body, can be modeled. The total bending energy (U_{Tb}) of the guide wire equals the sum of the bending energy present in each joint:

$$U_{Tb} = \sum_{i=0}^{k-1} U_b(\overline{x}_i) \tag{2}$$

No bending energy is defined for the tip joint (k) since only one segment is connected to this joint.

Guide wires are available in a variety of shapes. A distinction between completely straight guide wires and guide wires with an intrinsically curved tip can be made. In Equation 1, zero bending energy for joint i of a completely straight guide wire is achieved if two adjacent segments lie parallel:

$$(\overline{x}_i - \overline{x}_{i-1}) \;//\; (\overline{x}_{i+1} - \overline{x}_i) \tag{3}$$

The intrinsic curvature in curved tip guide wires is included in our model by associating a vector ($\overline{\omega}$) with each joint that represents a bias towards this curvature. The vector indicates the deviation from the straight line it would have formed with the previous segment if no intrinsic curvature would have been present. Zero bending energy for joint i belonging to an intrinsically curved part of the guide wire is then achieved if:

$$(\overline{x}_i - \overline{x}_{i-1} + \overline{\omega}_i) \;//\; (\overline{x}_{i+1} - \overline{x}_i) \tag{4}$$

Generally, when one side of a longitudinal object is subjected to a torque (τ), while resistive forces are present at the other side of the object, the object will show some torsion ($\varphi_{torsion}$). The amount of torsion depends on the torsion constant ($\kappa_{torsion}$) of the object:

$$\tau = \kappa_{torsion}\varphi_{torsion} \tag{5}$$

In practice, with the tip as an exception, guide wires have excellent torque control. In the current version of our model, the assumption has been made that guide wires have ideal torque control, i.e., the torsion constant of a guide wire approaches infinity.

To model the interaction between the guide wire and the vasculature, the outer hull of the guide wire is modeled by a set of points. This allows for modeling the non–zero thickness of guide wires (Figure 1).

2.2 Modeling the Vasculature

To model and test the motion and deformation of a guide wire, the vasculature in which the guide wire moves, needs to be modeled. Hereto we segmented 3D X-Ray data sets of phantom data. For the segmentation we used the fuzzy connectedness algorithm introduced by Udupa et al. [8]. The segmentations form the basis for modeling the vasculature.

The deformation properties of the vessel wall are determined by the tissue characteristics of the vessel wall and the tissue types surrounding the vessel. Incorporating all these aspects will most likely give the most realistic representation of the deformation of the vessel wall. For now, a simplified model is used for the vasculature and the vessel wall energy. Hookes' law [2] is used, which states that the deformation of an elastic material is proportional to the applied stress up to a certain point, called the elastic limit, beyond which additional stress will deform it permanently. Mathematically Hookes' law can be described as follows:

$$F = kd, \quad U = \frac{1}{2}kd^2 \tag{6}$$

Where F is the applied force, k the spring constant and d the deformation of the elastic body subjected to the force F. Integrating this formula with respect to d gives us an expression for the energy (U). The change in energy ($\delta U = \delta\overline{x} \cdot \frac{\partial U}{\partial \overline{x}}$) due to a displacement ($\delta\overline{x}$) can be expressed in terms of the gradient ($\delta U = \nabla U \cdot \delta\overline{x}$) of the energy at the new arrived position ((x, y, z)), where $\nabla U = (\frac{\partial U}{\partial x}, \frac{\partial U}{\partial y}, \frac{\partial U}{\partial z})$.

The energy formula in equation 6 is used to define a potential distribution which represents the elastic vessel wall energy expressed in gradients. Within the lumen of the vessel, the potential distribution equals zero. Elsewhere, the potential distribution has non–zero values according to the energy formula mentioned above, showing increasing values when moving further away from the vessel wall.

The total vessel wall energy (U_{Tvw}) the guide wire is subjected to equals the sum of the wall energy every single joint is subjected to (U_{vw}):

$$U_{Tvw} = \sum_{i=1}^{k} U_{vw}(\overline{x}_i) \tag{7}$$

The first joint (0) has no contact with the vessel wall and is therefore not considered. The vessel wall energy for every joint is the average of the vessel wall energy present in the sampled points on the outer hull of the guide wire.

3 Relaxation Algorithm: Minimization of the Total Energy

The developed algorithm, based on quasi-static mechanics, handles the relaxation of a guide wire within a vascular system after a forced translation or rotation of the proximal end of the guide wire has taken place. The determination of the input for the relaxation algorithm, i.e. the initial configuration of the guide wire after a forced translation or rotation, is explained in Section 4.

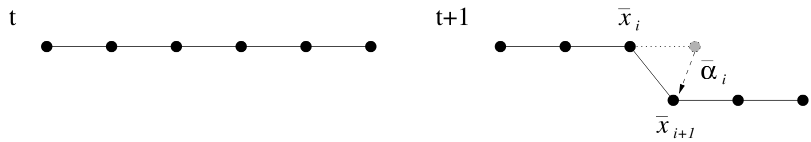

Fig. 2. A change in $\overline{\alpha}_i$ has influence on the bending energy of joints i and $i+1$ and on the positions of joints $i+1$ to k.

The relaxation process can be defined as finding a new configuration of joint positions on basis of the minimization of the total energy of the guide wire.

The bending energy of the guide wire and the vessel wall energy of the vasculature are both functions of the positions of all joints. Since a displacement of any joint will entail displacements of other joints as well, there is a high degree of interdependence between the joint positions in the total energy function. Therefore, a different set of variables which uniquely describe the positions of the joints is introduced:

$$\overline{\xi}_i = \overline{x}'_i - \overline{x}_i, \quad \overline{\alpha}_i = \overline{\xi}_{i+1} - \overline{\xi}_i \tag{8}$$

The vector $\overline{\xi}_i$ represents the difference between the new position of joint i ($\overline{x}'_i$) and the previous position ($\overline{x}_i$), i.e. the movement of joint i. The vector $\overline{\alpha}_i$ denotes the difference between the movements of two adjacent joints (Figure 2). A change in $\overline{\alpha}_i$ affects the positions of joints $i+1$ to the tip, but solely the bending energy of joint i and $i+1$. This constitutes a significant reduction of the interdependency of the variables and hence of the complexity of the algorithm.

After a forced translation over a fixed interval ($\overline{\xi}_0$) or a rotation ($\overline{\xi}_0 = 0$) of the proximal end of the guide wire, the guide wire responds to a translation $\overline{\xi}_i$ at each joint i that may deviate from the fixed translation $\overline{\xi}_0$ according to:

$$\overline{\xi}_i = \overline{\xi}_0 + \sum_{j=0}^{i-1} \overline{\alpha}_j \tag{9}$$

The purpose of the algorithm can now be defined as finding a specific set of $\overline{\alpha}$ vectors such that the resulting $\overline{\xi}_i$ corresponds to the situation in which the total energy is minimal. Since the distance between two adjacent joints is constant, the 3D $\overline{\alpha}$ vectors can be represented using a 2D parametrization, *viz.* an angle ψ_i around $\overline{\lambda}_i$ ($\overline{x}_{i+1} - \overline{x}_i$) and a scalar a_i which represents the length of $\overline{\alpha}_i$.

Minimizing the sum of the vessel wall energy and the bending energy, can be performed by finding the set of a_i values and ψ_i values for which the total energy is an extremum. This gives an analytical expression for ψ_i and a_i:

$$\begin{aligned} a_i &= \tfrac{|\tilde{G} - c_i \tilde{\Upsilon}|\lambda}{2G^P - c_i \varrho} \\ \psi_i &= \beta_i + \pi \end{aligned} \tag{10}$$

For a full derivation we refer the interested reader to a technical report [5]. Here, we will only intuitively describe the different terms. The terms basically represent trade–offs between relevant gradients of the vessel wall energy (which can be interpreted as forces acting on the guide wire) and the bending energy associated with joint i. $\tilde{G}$ denotes the transversal component of the sum of the gradients of the vessel wall energy at all joints from $i+1$ to the tip, and G^P expresses the longitudinal component of this term. The longitudinal component of the bending energy associated with joint i is represented by the term $c_i \varrho$, whereas $c_i \tilde{\Upsilon}$ denotes the transversal component of this energy. In equation 10, β_i is the angle expressed in polar coordinates represented by the vector $\tilde{G} - c_i \tilde{\Upsilon}$, which denotes the trade–off in the transversal plane between the relevant vessel wall energy gradients and the bending energy associated with joint i. The solutions β_i and $\beta_i + \pi$ denote the values for ψ_i in which ψ_i represents an extremum. The minimum energy state is obtained when ψ_i equals $\beta_i + \pi$. Clearly, when an equilibrium between the local bending energy and the relevant vessel wall gradients has been achieved, i.e. $\tilde{G} = c_i \tilde{\Upsilon}$, the expression for a_i equals zero, which means that the i–th joint is in a steady state.

4 Initialization of the Relaxation Algorithm

In the following subsections it is explained how after a forced translation or rotation of the guide wire, an initial configuration of the guide wire is calculated that can serve as a starting point for the relaxation algorithm.

4.1 Translation

Given the current position of the guide wire $(\overline{x}_0, \ldots, \overline{x}_k)$ and a change of $\overline{x}_0$ as a result of a forced translation of the proximal guide wire body into the introducer sheath, the initial configuration of the guide wire is calculated by adding the movement of joint 0 ($\overline{\xi}_0$) to every single joint position, see Figure 3.

4.2 Rotation

Given the current position of the guide wire $(\overline{x}_0, \ldots, \overline{x}_k)$ and a forced rotation of the proximal end of the guide wire, the initial configuration of the guide wire is calculated not by translating the joints, but by rotating local (P, Q, S) coordinate systems that are associated with each joint and initialized as follows:

$$\hat{e}_i^P = \frac{1}{\lambda} \overline{\lambda}_i, \quad \hat{e}_i^Q = \hat{n}_{\mathcal{L}}, \quad \hat{e}_i^S = \hat{e}_i^P \times \hat{e}_i^Q \tag{11}$$

Where $\hat{n}_{\mathcal{L}}$ is the unit vector perpendicular to plane $\mathcal{L}$ containing all joints under the circumstances that the guide wire is outside the vasculature and no external forces are applied to it.

By rotating the local coordinate systems, the local characteristics associated with each joint are rotated as well, like the $\overline{\omega}$ vector (Figure 3). After rotation, the local semantics of each $\overline{\omega}_i$ remains identical, but globally the orientation has changed and thus the joints connected to joint i desire a new position in order to ensure that the internal energy stays minimized (Equation 4). The new desired configuration of the guide wire will be found when this initial configuration is given to the relaxation algorithm.

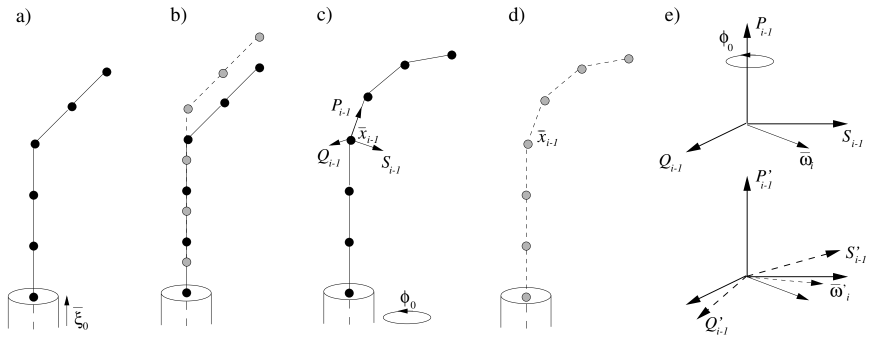

Fig. 3. Determining the initial configuration after a forced translation or rotation, that will serve as input for the relaxation algorithm: *a,c*) current guide wire configuration, *b*) the initial configuration after a forced translation, *d*) the initial configuration after a forced rotation. *e*) For clarity only one local coordinate system has been visualized. After forced rotation, the global semantics of the $\overline{\omega}$ vectors has changed. In the relaxation process, the guide wire will deform so as to minimize the total energy, taking into account the preference for the predefined intrinsic curvature.

5 Experiments

To demonstrate the performance of the relaxation algorithm used in the simulation of the propagation of a guide wire, 3D X-Ray data from an intra–cranial anthropomorphic vascular phantom, including the circle of Willis and an aneurysm (Figure 4a), have been acquired. Phantom data are very useful since during interventions in patients only projection images are acquired. In phantom experiments, 3D X-Ray images can be acquired, providing the advantage that the actual position of the guide wire in experiments can be visualized for comparison with the simulation.

To obtain a realistic simulation, the correct vessel wall elasticity and guide wire characteristics (spring constants, $\overline{\omega}$'s) need to be incorporated in the model.

During an experiment, a guide wire with a curved tip is propagated in the intra–cranial vascular phantom. A time sequence of 30 3D X-Ray images have been acquired during this experiment. The position of the guide wire is also modeled for different insertion depths. Figure 4c shows the position of the guide wire inside the phantom after propagation of the guide wire until it has reached the aneurysm. Figure 4b represents the simulation of the propagation of a guide wire until it has reached the same position. Comparing Figure 4b and 4c, a large similarity can be detected, indicating a plausible behavior of the simulation.

6 Conclusions

An algorithm for simulating the propagation (translation and rotation) of a guide wire for intravascular interventions has been presented. An analytical solution to the minimization of the energy is used in this algorithm. The algorithm is based on quasi-static mechanics. The current discretized guide wire representation we introduced, allows to model guide wires with different physical characteristics

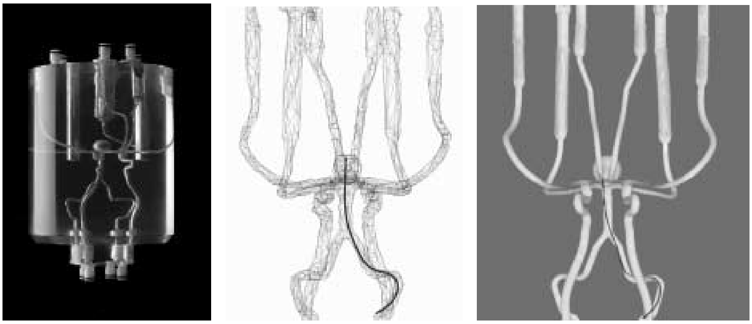

Fig. 4. *a*) An intra–cranial anthropomorphic vascular phantom including the circle of Willis and an aneurysm. Visualization of a simulated (*b*) and a real (*c*) propagation of a guide wire in the intra–cranial phantom until it reaches the aneurysm.

by changing the segment length, defining different values for the constants for the bending energy per joint and by adjusting the $\overline{\omega}$ vector per joint. Results indicate plausible behavior of guide wire propagation in a phantom data set, where the actual position could be validated visually using 3D X-Ray imaging. This verifies that the methodology we proposed, based on a relaxation algorithm, gives realistic results. However, for a more realistic modeling of motion and forces in patients, additional information needs to be incorporated. This can readily be achieved with the proposed paradigm. Furthermore, for extensive validation a simulation device is currently being constructed which allows controlled translation and rotation of the guide wire in experiments.

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